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(71) Applicant *(for all designated States except US)*: PAIEON, INC. [US/US]; 747 3rd Avenue, New York, NY 10017-2803 (US).

(72) Inventors; and

(75) Inventors/Applicants *(for US only)*: KLAIMAN, Moshe [IL/IL]; Ha'rava 21/2 Street, 70700 Gedera (IL). ZARKH, Michael [IL/IL]; 9 Oranim Street, 54052 Giv'at Shmuel (IL).

(74) Agent: HOPKINS, Brian, P.; Mintz, Levin, Cohn, Ferris, Glovsky and Popeo, P.C., 666 Third Avenue, New York, NY 10017 (US).

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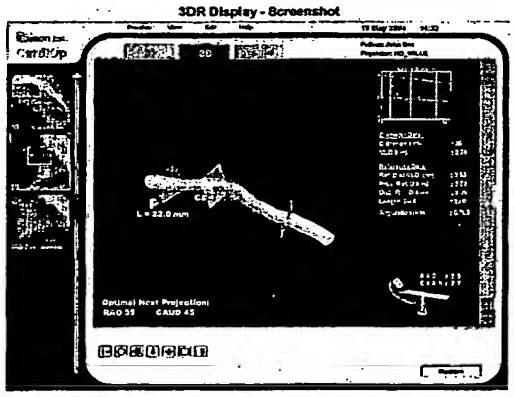


(57) Abstract: Embodiments of the present invention include methods and systems for three-dimensional reconstruction of a tubular organ (for example, coronary artery) using a plurality of two-dimensional images. Some of the embodiments may include displaying a first image of a vascular network, receiving input for identifying on the first image a vessel of interest, tracing the edges of the vessel of interest including eliminating false edges of objects visually adjacent to the vessel of interest, determining substantially precise radius and densitometry values along the vessel, displaying at least a second image of the vascular network, receiving input for identifying on the second image the vessel of interest, tracing the edges of the vessel of interest in the second image, including eliminating false edges of objects visually adjacent to the vessel of interest, determining substantially precise radius and densitometry values along the vessel in the second image, determining a three dimensional reconstruction of the vessel of interest and determining fused area (cross-section) measurements along the vessel and computing and presenting quantitative measurements, including, but not limited to, true length, percent narrowing (diameter and area), and the like.

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# SYSTEM AND METHOD FOR THREE-DIMENSIONAL RECONSTRUCTION OF A TUBULAR ORGAN

## CLAIMS TO PRIORITY AND RELATED APPLICATIONS

The present application claims priority under 35 U.S.C. §119(e) of U.S. provisional patent application nos. 60/505,430, filed September 25, 2003, 60/506,178, filed September 29, 2003, and 60/577,981, filed June 7, 2004, each disclosure of which, in its entirety, is incorporated herein by reference.

## BACKGROUND OF THE INVENTION

### Field Of The Invention

The present invention relates to medical imaging systems, and more specifically to medical imaging systems for use in angiography, for example.

### Background Of The Invention

A stenosis in a blood vessel, for example, an artery refers to narrowing of the artery lumen due to plaque formation on the interior wall of the artery. The severity of the narrowing depends upon the amount of cross-sectional area of the lumen that is occluded by plaque. While narrowing of the arteries may occur in any artery of the body (e.g., carotid arteries), particular concern has been placed on investigating the narrowing of arteries of the heart, the coronary arteries (coronary heart disease), since narrowing of these arteries is one of the primary causes of heart attacks. Accordingly, coronary angiography refers to the process (and associated systems) of investigating coronary

2D QCA systems usually provide diameter based analysis of the lesion and not densitometry analysis. In some cases, densitometry analysis is provided via the usage of DSA, but not for scenes that include motion, like the coronaries.

The 3D QCA methods generally include the following steps: image acquisition, vessel extraction from the 2D projections. The 3D QCA systems additionally include imaging geometry recovery, point-by-point matching (between images) and, of course, 3DR. The QCA of the 3D system generally includes, morphology assessment (including vessel foreshortening, overlapping, angulation, tortuosity), and in some cases measurements, usually true length and diameter information. However, cross-section area measurements are rarely addressed, although attempts have been made to achieve a precise representation of cross section profile along the vessels. A method based on some heuristics in a framework of algebraic reconstruction approach was suggested in U.S. patent no. 6,301,498 to Greenberg. However this method requires a special arrangement of at least four (4) acquisitions from different directions orthogonal to the artery.

Also, in both 2D and 3D QCA systems and methods, one important aspect of measurements and stenosis severity is the establishment of healthy vessel measurements. Systems and methods that present healthy vessel (or related) measurements use, for example, the interpolation of values based on measured diameters at proximal and at distal portions. This step is critical, since it is a basis for many measurements. At the same time, this step is very sensitive and could easily produce incorrect measurements.

Other problems exist with reference to the methods for the existing 3D imaging systems. For example, with image acquisition, prior art systems utilize either bi-plane acquisition, rotational acquisition, or single projection (image) acquisition (the most general approach (see U.S. patent nos. 6,047,080 and 6,169,917). Although bi-plane acquisition minimizes distortions due to cardiac cycle phase, the technique is insufficient in some situations of epi-polar geometry ambiguity. With regard to rotational acquisition systems, although close in time, these systems do not solve either a cardiac phase problem or the epi-polar geometry ambiguity.

With regard to imaging geometry recovery, the number of control points needed for geometry recovery depends on the type of transformation that is found and

point on one image, the search for the corresponding point on another image may be found on the epi-polar line (intersection of the epipolar plane with the image plane). However, this approach yields sufficient results only if: (i) the imaging geometry model adequately relates the organ and its 2D image, and (ii) the imaged vessel does not change its shape between the image acquisitions. This is why, in clinical practice, the restrictions of the straightforward epi-polar geometry approach are very limiting in terms of accuracy and quality of the 3D model.

In view of the above-mentioned short comings of the prior art, current 2D QCA systems do not deliver sufficient support for coronary angiography (for example) and current 3D QCA systems are not in clinical use since these systems either deliver incorrect results or are too cumbersome to use.

Thus, there exists a need for a 3DR system which can be used in clinical procedures (e.g., angiography) that delivers a system that may include a practical, intuitive, easy-to-use, robust solution to overcome at least one and preferably all of the above-mentioned disadvantages of the prior art systems and methods.

## SUMMARY OF THE INVENTION

Accordingly, embodiments of the present invention overcome the drawbacks and problems associated with the prior art systems, and present simple to use and straight forward systems and methods for accurately imaging and creating a 3DR of a tubular organ which may be used with conventional X-ray angiography systems. Specifically, some embodiments of the present invention present methods and systems for 3DR of a single vascular structure of interest, using two (and in some embodiments, more than two), 2D X-ray images.

Briefly, some embodiments may include one or more (and, in some embodiments all) of the following: acquisition of cine-runs, projection angulation and ECG information (e.g., via Analog and/or DICOM), system calibration to process images (e.g., catheter calibration), marking of two or more images, edge tracing, with pre and post processing to eliminate potential incorrect distortions of the edge, detection of centerline,

The determination of the fused area may include determining a plurality of healthy diameters (and preferably all healthy diameters) along the vessel of interest to be used as a physical reference, normalizing a majority of the data (and preferably substantially all the data, and most preferably, all the data), e.g., diameters and cross-section values into physical units, using the physical reference, fusing a majority of the data (preferably all or substantially all) into single area measurements and weighting each source of data according to the reliability of the data. The weighting may be computed as a function of the views geometry and/or 3D vessel geometry.

The input for identifying the vessel of interest may include three points: a first point to mark the stenosis general location, a second point proximal to the stenosis, and a third point distal to the stenosis.

However, the input may also comprises markers for two (2) points for at least one of the first and second images, where one of the two points is anywhere proximal to the stenosis and the other point is anywhere distal to the stenosis. The markers may also comprise two (2) points for the first image and one (1) point for the second image, where one of the two points is anywhere proximal to the stenosis and the other point is anywhere distal to the stenosis and one point may be an anchor point identified automatically on the first image.

The elimination of false edges may comprise detecting of one or more "bubbles" (see description below) adjacent the vessel of interest. A novel embodiment for detecting such bubbles (e.g., false edges) may include defining a region of interest substantially parallel to a primary centerline, detecting at least one cluster of pixel data, adjacent to the vessel of interest, wherein each cluster of pixel data having a predetermined brightness level greater than a brightness level of surrounding pixel data, selecting an arbitrary pixel within each cluster, selecting a second pixel provided on a lane bounding the region of interest for each arbitrary pixel of each cluster, and establishing a barrier line to define an edge for the vessel of interest by connecting a plurality of arbitrary pixels with a corresponding second pixel. Upon the tracing each edge of the vessel of interest, the traced edge avoids each barrier line.

substantially parallel to at least one edge of the vessel of interest, establishing a parametric grid covering the vessel of interest and a neighboring region, where the parametric grid includes a first parameter of the vessel of interest along the length thereof and a second parameter for controlling a cross-wise change of the vessel of interest and sampling the image using the grid to obtain a plurality of corresponding gray values - the gray values are investigated as functions on the profile lines. The method may also include substantially eliminating detected occluding structures on the outside of the vessel of interest, the structures being detected as prominent minima of the parameters, substantially eliminating prominent minima detected on the inside of the vessel of interest, averaging gray values in a direction across the vessel of interest separately for each side of the vessel of interest, determining a linear background estimation on the grid inside the vessel of interest and determining cross-sectional area using the eliminated prominent minima.

Embodiments of the invention may include determining healthy vessel dimensions using an iterative regression over a healthy portion of the vessel of interest. In particular, iteration comprises a compromise between a pre-defined slope and a line that follows healthy data. In one embodiment, the compromise is toward the line that follows the healthy data if the line corresponds to actual data over a plurality of clusters. The determined healthy dimensions of the vessel of interest may be displayed, either in 2D and/or in 3D.

Three-dimensional reconstruction of the vessel of interest may include: determining a conventional epi-polar distance $p_1$ for the plurality of centerline points in the first image, determining a conventional epi-polar distance $p_2$ for the plurality of centerline points in the second image and re-determining $p_2$ substantially in accordance with $p_{2new} = p_2 + \delta$, where $\delta$ is a smooth compensatory function establishing correspondence of one or more landmark points.

An epi-polar indicator, and associated means (e.g., application program/computer instructions for a processor), may be included with various embodiments of the present invention. Accordingly, after receiving input for identifying the vessel of interest in the second image, the epi-polar indicator may be displayed for indicating a concurrence

densitometry values along the vessel of interest in the second image, determining a three dimensional reconstruction of the vessel of interest, determining fused area (cross-section) measurements along the vessel and establishing the 3DR of the vessel of interest.

Other embodiments of the present invention may include a system for three-dimensional reconstruction (3DR) of a single blood vessel using a plurality of two-dimensional images is provided (according to any of the foregoing, for example) and may also include an angiography system comprising a platform for scanning a patient, a C-ARM X-ray system including an x-ray source, a detector, a step motor for moving the C-ARM, and a workstation for doing QCA. The workstation may include display means for displaying a first image of a vascular network, and a second image of the vascular network and the 3DR, input means for identifying a vessel of interest on the first image and the second image, tracing means for tracing the edges of the vessel of interest in each image including elimination means for eliminating false edges of objects visually adjacent to the vessel of interest in each image and a processor.

Still other embodiments of the invention are directed to computer readable media (e.g., floppy discs, hard-drives, CDs, DVDs, smart media and other flash storage), whether permanent or temporary, for storing one or more application programs made up of computer instructions (or just computer instructions) for enabling a computer (e.g., processor, and/or a workstation/network) to perform the methods according to the various embodiments of the present invention.

Any of the embodiments of the invention may also be used with existing angiography systems, or other vessel imaging systems. The relation of the present invention to such systems is readily apparent to one of ordinary skill in the art in view of the present disclosure.

Other embodiments, as well as objects and advantages of the present invention will become more clear with reference to the following detailed description and attached figures as briefly described below.

Fig. 18 is a schematic illustrating a principle of densitometry according to some embodiments of the present invention.

Fig. 19A is an image illustrating profile lines of a vessel of interest for computing densitometry.

Fig. 19B is a graph of densitometry values associated with the image of Fig. 19A.

Fig. 20A and 20B represent first and second images of a vascular network for illustrating point-to-point matching.

Fig. 21A is an image of a stenotic vessel for illustration of healthy artery computation.

Fig. 21B is a graph illustrating healthy artery computation of the stenotic vessel of Fig. 21A.

Fig. 22A is another image of a stenotic vessel for further illustration of healthy artery computation.

Fig. 22B is a graph illustrating the healthy artery computation of the stenotic vessel of Fig. 22A.

Figs. 23-28 are images of a stenotic vessel of interest, with reference to determining a healthy display of the vessel.

Fig. 29 is a screenshot for a 3DR system according to the present invention illustrating a 2D image related display (including healthy artery display in 2D).

Fig. 30 is a 3DR of a vessel of interest.

Fig. 31 is a screenshot of a 3DR, including display of information associated with the 3DR.

Fig. 32 illustrates an example of a pop-list that appears on screenshots of a system according to the present invention (also illustrating a 3DR of a vessel of interest).

7 is capable of rotating the radiation source and the radiation detector about the Z-axis, which is the longitudinal axis of the patient's body, and also about the X-axis, which defines, with the Z-axis, the plane of the horizontal body support.

The electronics which may be included with the system of Fig. 1 may include an angiography system controller 10 which controls the radiation source and also the step motor to successively produce the exposures of the body from a plurality of different angular positions with respect to the body. The controller may also receive the electronic outputs from the radiation detector elements in the CCD camera. A computer work station 11 may be included which controls the angiography system controller 10 to produce a two-dimensional images 12 of blood vessels projection to any selected play (cine-runs), as well as 3DR images 13. Control is preferably synchronized with a cardiac and/or respiratory gating signals produced by an ECG sensor and/or a respiration sensor (not shown), so that images of the blood vessels may be obtained during the same point during a cardiac cycle or respiration cycle.

The workstation may include the application programs and/or hardware for enabling the operation of the systems and methods of the embodiments of the invention for 2D and 3DR, as well as the associated QCA. Also, the systems and methods according to embodiments of the present invention may be and add-on component to the above-described configuration for a catheterization room. In some embodiments, another workstation, including hardware and software, may be interfaced to the catheterization room, for receiving cine-runs, and optionally, C-ARM angulation and ECG to process and present 3DR.

## FIRST GROUP OF EMBODIMENTS

Image Acquisition

Two-dimensional (2D) X-ray images of a plurality of cine-angio runs are captured and presented on the monitor substantially in real-time during catheterization of a patient. In addition to the images, C-arm angulation data and ECG data may also be acquired. Using the ECG sensor, an ECG Gating process may be used to present the optimal

Edge Detection (edge tracing)

Initially, a primary centerline may be extracted using known algorithms (such as dijkstra optimization or wave propagation method). The only property the primary centerline should possess is that it be a path inside the marked vessel. In this regard, the user marking points, which can be located outside the vessel due to an imprecise user's pointing, may be automatically checked and moved, if necessary, into the vessel. Accordingly, the tracing algorithm may use these properly located marking points to an extract a primary centerline.

For each image, the edges of the marked vessel of interest are traced. Although edge detection (i.e., edge tracing) may be accomplished via known methods using known algorithms (see, for example, Gradient Field Transform, "A New Approach For The Quantification of Complex Lesion Morphology: The Gradient Field Transform;...", Zweit & Reiber, JACC vol. 24; "Single Source Shortest Path"; Introduction To Algorithms; Cormen, Leiserson & Riverst., p. 527; each of which is herein incorporated by reference in its entirety). However, using these known methods, edge detection in angiography poses many difficulties, for which embodiments of the present invention address.

Such difficulties relate to a detected edge of a vessel of interest "detouring" off the actual edge of the vessel onto an edge of a visually adjacent vessel (or other feature/object) from the complex vascular structure which may surround the vessel of interest (in which the vessel of interest may be apart of)(see Figs. 3-4, illustrating a complex network of vessels and an incorrect edge trace 410). Moreover, a similar phenomenon is recognized where the end-point of the vessel of interest is marked, near which likes another parallel (or substantially parallel) vessel. Accordingly, prior to detecting the edges of the vessel of interest (using, for example, the above edge detecting methodologies, or modified versions thereof), embodiments of the invention conduct pre-processing to substantially reduce and preferably eliminate such detours from appearing as the final edge.

The phenomenon is shown in Fig. 5 and is addressed by seeking out what is referred to as a "bubble" 510, adjacent the vessel of interest, which results in an incorrect

Accordingly, after a primary edge 1200 (Fig. 12) is found, bumps are sought out. Starting from a point on the edge 1300, the closest point on the primary centerline (or opposite edge or any line substantially parallel to the vessel) is found and a distance between the two is found (arrow 1310). Then, from a point on the centerline, the closest point on the edge is found and a distance between the two is found, which is denoted by arrow 1320. Deviation from the centerline is defined as an absolute difference between distances 1320 and 1310.

Preferably, all edge points are checked for as being bump points. Thereafter, for every point on the primary edge, a combined function is calculated. This function aggregates two components: deviation from centerline and gradient cost function (for example, a condition on the gradient value can be expressed via a gradient cost function which may be inversely proportional to the gradient magnitude). Specifically, a suspected bump point having a big deviation from centerline and/or low gradient, may be considered an actual bump point. The combined function, in particular, can be product of deviation from centerline and gradient magnitude. A bump comprises a plurality of bump points.

The detected bumps are corrected by "cutting" the bumps from the edge. After bump points have been determined (which may include one or more neighboring edge points), an area 1405 of the bump is then determined, using the outer border of the bump 1400 and a cutting line 1410 as inner border. The appropriate cutting line is finally determined by a line which maximizes the ratio between the bump area and a function of cutting line length, for example, a power of the cutting line length, and which is also the correct edge of the vessel of interest. This "cuts" the bump from the imaging of the vessel and establishes the correct edge of the vessel.

Centerline Definition

The centerline definition, being an input for determining radius and densitometry values, ultimately determines stenosis measurements, and thus is very important. By definition, the centerline is a line passing inside the vessel, between the edges. Every point in the centerline should be equally distant from the edges (i.e., the center). This is

prior art methods regarding DSA (Digital Subtraction Angiography), which are very useful for static objects, but are hard to implement for a moving coronary vessel. Thus, other described methods are trying other approaches to "subtract" the background; these methods are very problematic, since they are very local (see Fig. 17). Specifically, as shown, dashed line 1710 represents a centerline of the vessel in interest. As briefly mentioned above, the classical approach to compute densitometry is to compute the background gray-levels along segments perpendicular to the centerline (for example, black lines 1720, 1730) and to "subtract" those values of the background (e.g., outside vessel boundaries/edges) from the vessel's gray-value. If indeed the perpendicular segments pass through background that is common to the artery (for example, the left segment passes through the catheter), such a method may work. The vessel of interest also "goes over" the catheter, thus, subtracting the catheter gray-level values is justified.

On the other hand, if the right segment 1730 goes through a branching vessel; the gray level values for the vessel of interest along this segment are not influenced by the branching vessel (unlike the previous example of the catheter). Thus, it is erroneous to subtract the "background" (actually the branching vessel's) gray-values from those of the vessel of interest.

Accordingly, one embodiment of the invention presents a novel algorithm to "subtract" the background influence in a vessel. Initially, profile lines 1810 (Fig. 18) along the background are drawn, parallel to the edge. This way, the background analysis is much more global and may account for many things the classic approach cannot.

In order to consistently evaluate the background, a two parametric grid covering the vessel and neighboring region are applied. One parameter controls the change of the vessel along its length and the second parameter controls the change of the vessel cross-wise. The image is then sampled on the grid. Obtained gray values are investigated as functions on the lines parallel to the vessel (lines 1810, Fig. 18). The crossing vessels and other occluding structures are detected as prominent minima of the functions and preferably eliminated. The similar minima elimination is also performed on the grid inside the artery. The values of the grid outside the vessel are averaged in a direction across the vessel on both sides separately, and linear background estimation is calculated

points in order to improve reconstruction process. Specifically, in the framework of an orthographic projection, the epi-polar principle prescribes the corresponding points to be in equal distance (epi-polar distance p; see Figs. 20A-20B) to a reference epi-polar line. Reference points can be marked by an operator in all images or a reference point marked by the operator in one image may then be refined in order for it to be accurately located by a local correlation algorithm (for example) in other images or the reference points can be found automatically in all the images.

The following types of landmark feature points may be utilized for calculation of improved epi-polar distance on an image: branching points (B); prominent features of diameter function (C1,C2); local extremes of epi-polar distance (D) as a function of centerline point; and points of extreme curvatures (E).

A vessel's centerline points are preferably matched according to the match of improved epi-polar distances. Specifically, a conventional epi-polar distance p is calculated for artery centerline points of reference image $p_1$ in Fig. 20A and for artery centerline of the second image $p_2$ in Fig. 20B. Then the second epi-polar distances $p_2$ are re-calculated in a form $p_{2new} = p_2 + \delta$ in order to provide equal epi-polar distances at landmark points, where $\delta$ may be a smooth compensatory function establishing correspondence of the landmark points. If $p_1(LM)$ and $p_2(LM)$ are the epi-polar distances of a landmark point, then the compensatory function includes a value $\delta(LM) = p_1(LM) - p_2(LM)$ at this landmark point. See illustration of value $\delta$ for landmark point E. It is worthwhile to note that the compensatory function $\delta$ is calculated per specific vessel. This approach has a straightforward extension for the case of reconstruction from three (3) images. The two compensatory functions for the second and third images $\delta_2$ and $\delta_3$ have values $\delta_2(LM) = p_1(LM) - p_2(LM)$, and $\delta_3(LM) = p_1(LM) - p_3(LM)$ at landmark points.

Healthy Artery Computation

Embodiments of the present invention obtain a graph of measures: diameter or cross-section area, along the artery. In order to perform lesion analysis to compute

These improvements are significant to the classic method, and provide not only better and more robust results, but also enable a system to consider more complicated cases, such as ostial lesions (which are lesions without a proximal or distal healthy portion for the vessel).

For example, Fig. 21A illustrates an example for a "normal" stenotic vessel, with both proximal 2110 and distal 2120 healthy portions. In a representing iteration of the healthy artery computations, Fig. 21B, there will be two clusters of points: one in the proximal part and one in the distal part (the bulleted points in the figure), for which the values of the radius line (2130) is relatively close to the values of the "regression" line (2140). Since there are two clusters distributed along the vessel, the new line (which strives to get closer to the data) will be accepted (rather than striving to stay closer to the predefined slope (2150).

Figs. 22A-22B represent another example. In this example, however, the vessel of interest presents an ostial lesion (or defused disease). As can be seen, the vessel has a healthy proximal portion 2210, but is stenotic through all its distal part 2220. This, of-course, is manifested in the radius graph of Fig. 22B. In this case, the "regression line" 2230, includes one cluster of points, in which the radius value is close to the regression value. Thus, in this case, the result of the iteration will be closer to the default slope 2240 than to the regression line.

It is worth noting, that this step of healthy artery computation is described in two contexts, computation and for 2D display (see below). Accordingly, the above computation is preferably performed first, and then it serves as an input for the 2D display procedure. The difference between the two steps is that the computation step is generally related to the healthy values, while the second step (of display) may also be related to the "symmetry" of this values versus the lesion (for example, how to locate a value of 5mm healthy "around" a 3mm lesion).

healthy radius and this healthy radius in turn is greater than the input radius at this point, a new couple of anchor points are found.

One point of the new couple is the new point. The second point constituting the new couple is determined via healthy radius and a point from the opposite edge corresponding to the new one. Namely, the second point constituting the new couple lies in the twice healthy-radius distance on the straight line connecting the new point with its counterpart. If the deviation is greater than the correspondent healthy radius and this healthy radius is less than the input radius at this point (e.g., an aneurysm) then a new couple of the anchor points also exists. The points of the new couple lie on the straight line connecting the new point and a point from the opposite edge corresponding to the new one. The distance between the points of new couple, as in the previous case, is equal to twice healthy-radius. But, contrary to the previous case, the points of the new couple are located symmetric relative to corresponding centerline point. A result of the termination of recursion is a list of anchor points. The healthy edge is finalized via interpolation between anchor points (for example, spline interpolation). See Fig. 29 showing, at center, two-dimensional, healthy artery display.

Three-Dimensional Healthy Artery Display

The same notion for 2D of a healthy vessel is applied for 3D. As shown in Figs. 30-32, transparent area 3010 is visualizing an approximation of the healthy vessel. Similarly to the 3D vessel reconstruction, the healthy 3D artery is defined by 3D healthy centerline and 3D healthy diameters. For 3D healthy centerline calculation, the known point-by-point matching of 2D centerlines may be utilized, applying it to the healthy 2D centerline points nearest to the available matched pair. The 3D healthy diameters may then be taken as a diameter corresponding to the healthy (reference) diameter. The cross section area may be a result of the fusion algorithm described below and the healthy diameter is (iterative) regression line for sqrt(cross section area/$\pi$).

More specifically, the healthy line of the average diameter may be used as a reference line. In that regard, substantially all (and preferably all) data may be transformed (radius and densitometry per run) using (for example) the ratio between the data's healthy line and the reference healthy line.

*RadsNorm = RadAvReg / RadsReg * Rads,*
*RadensNorm = RadAvReg / RadensReg * Radens,*

*Where:*
*RadsNorm are Normalized Radius values,*
*RadensNorm are Normalized Densitometry-derived-Radius values,*
*RadAvReg are healthy (regression line) values derived from average radius graph,*
*RadsReg are healthy (regression line) values derived from specific radius graph,*
*Rads are specific radius graph values,*
*RadensReg are healthy (regression line) values derived from specific densitometry-derived-radius graph,*
*Radens are specific densitometry-derived-radius graph values,*

The fused area may be calculated as a weighted sum of densitometry areas and areas calculated via product of diameters (for example). The weights may be determined locally and may depend on viewing directions and/or local 3D centerline direction. The weight of densitometry area may be maximal if the corresponding view is orthogonal to the centerline direction, while weight of product of diameters may be maximal if both views are orthogonal to the centerline direction and in addition mutually orthogonal.

An ellipse area may be used to express the area as product of diameters and a circle area may be used to express the area as the power of the cross-section-derived diameters.

*Sellipse(i,j)=pi*RadsNorm(i)*RadsNorm(j), i != j, i,j=1,2,..., NumberOfViews*
*Scircle(k)=pi*RadensNorm^2, k=1,2,..., NumberOfViews.*

It is worth noting an additional consideration. In some embodiments, the above definition gives some priority to the area measurements originated from densitometry since $W(i,j) < W(k)$, for $k=i$ and $k=j$. While the elliptical area assumption may suffer from probable inconsistency, the area evaluation via densitometry does not bear such defect (as mentioned above), and the priority to the densitometry may be reasonable.

Limited Marking 3DR

While the above embodiments disclose (generally) the use of three (3) marking points per image from at least two different cine-angio runs, other embodiments of the invention may utilize less marking points. For example, in some embodiments, the operator may simply mark two (2) points per image for two cine-angio runs, or, in other embodiments, the operator may mark two (2) points for a first image from one cine-angio run and one (1) point for one or more additional images (from other cine-angio runs).

For example, an operator may mark two (2) points proximal and distal (to a stenosis) on one image of one cine-angio run. This run may be referred to as a "Master" run, and the selected corresponding image to a "Master" image. The system then calculates centerline and edges and a "stenosis" point on the artery (this point is not required to be the actual stenosis point, but, rather, a reference point). The operator then selects images from two additional runs (the "Slave" runs) and marks the location of the "stenosis" point on the images from the Slave runs. After reception of each stenosis point on the images from the Slave runs, the systems performs tracing on the image of the Slave runs, then presents the results of the trace and the 3DR. This reduction of markings may be enabled in embodiments of the present invention using path optimization algorithms including, for example, a generic dijkstra algorithm or a wave propagation algorithm (WPA) adopted for the image trace.

Thus, using a WPA, for example, after image, source point and target point are input, with respect to the Master image, a path is found connecting the source point and target point with minimal cost (e.g., sum of image gray levels to some power). For the images from the Slave runs, the target set is an epi-polar line, instead of point, and the result is a path connecting the source point ("stenosis" or anchor point) to the target line.

## SECOND GROUP OF EMBODIMENTS

It is an objective of this group of embodiments of the present invention to provide a method and system for three-dimensional reconstruction of a tubular organ from angiographic projections. Specifically, the second group of embodiments improve the epi-polar geometry approach for 3DR by providing additional considerations to the three-dimensional reconstruction process, thus providing accurate correspondences between different projections and thus providing an accurate 3D model even in the presence of the mentioned geometrical distortions and epi-polar problem condition.

The suggested reconstruction method, according to the second group of embodiments, is based on epi-polar geometry enhanced by integrating other considerations to the reconstruction process. These other considerations including, for example, the tubular organ's parameters derived from the image, such as Radius and Densitometry (Gray-level) values, along the tubular organ's centerline and local centerline directions. Other considerations, which are derived from the tubular organ's characteristics, can be also incorporated. The current group of embodiments provide a method for three-dimensional reconstruction from two two-dimensional angiographic images and a method for three-dimensional reconstruction from three or more two-dimensional angiographic images. These embodiments further provide a solution for three-dimensional reconstruction for the case where a common reference point between all two-dimensional images is given as well as for the case where the reference point is not given; in this case, a novel approach is provided for obtaining a reference point by means of correlating the invariant functions.

Accordingly, some of the embodiments according to this second group include a method for establishing correspondence between projections of a tubular organ visible in angiographic images comprises:

(a) extracting centerlines of the tubular organ on two angiographic images,

(b) computing features along centerlines points: radius of the tubular organ, centerline direction, projected cross section area of the tubular organ (densitometry); these features compose invariant functions, which are used to match between centerlines,

Epi-polar principle defines that, given two 2D projections, every point on the first image defines an epi-polar line on the second image (and vice a versa); the 2D point on the second image that corresponds to the 2D point on the first image is restricted to this epi-polar line.

Three-dimensional reconstruction using epi-polar geometry could be described as follows:

(a) given 2D centerlines in two projections, each 2D point in the first centerline defines an epi-polar line that intersects the centerline of the second image, this intersection point is the 2D point on the second image that corresponds to the 2D point on the first image,

(b) each of these 2D points defines a projective line (meaning a line from the source 3D point to this projected 2D point). Thus, the intersection of the two projective lines finds the corresponding 3D source point (Ideally two lines are intersecting, but in practice they do not, so criteria such as minimal distance point should be defined).

(c) the sequence of resultant 3D points is the three-dimensional reconstruction of the tubular organ.

This described process of three-dimensional reconstruction using epi-polar geometry has many insufficiencies. Accordingly, the second group of embodiments brings answers to these insufficiencies based on exploitation of additional invariants apart from epi-polar distance for obtaining accurate match between 2D projections of a tubular organ and 3D reconstruction. One invariant is the radius function behavior along the projected arteries. In the invention, a general relation also is established between projected density of the arteries and epi-polar geometry. The relation allows calculation

(2) $D_i=D-(V_i^TD)V_i$ , i=1,2.

Note, the vectors $D_1$ and $D_2$ in (2) are not normalized.

Denote by $V_{12}$ the unit vector orthogonal to two view vectors $V_{12}= V_1\times V_2/|V_1\times V_2|$. The vector $V_{12}$ is the vector orthogonal to epi-polar planes of the two images. The measure of the projected orientation of the tubular organ relative to epi-polar plane E is, by definition, a scalar product of $V_{12}$ and the tubular organ's direction:

(3) $E_1\equiv D_1^T V_{12}/|D_1|$ and $E_2\equiv D_2^T V_{12}/|D_2|$.

Theorem: the ratio of projected area and the visible epi-polar orientation is invariant for every pear of views, i.e.

(4) $S_1/E_1= S_2/E_2$.

Proof:

Using (1) and (2), we obtain:

(5) $S_1^2(1-(V_1^TD)^2)= S_2^2(1-(V_2^TD)^2)$

(6) $D_1+(V_1^TD)V_1= D_2+(V_2^TD)V_2$ .

Multiplying (6) by $V_{12}$ we obtain $D_1^T V_{12}= D_2^T V_{12}$ and using notation (3)

(7) $|D_1| E_1=|D_2| E_2$.

Method for 3D reconstruction from 2D projections

For simplicity sake, the process will be described for two 2D projections. The process for three or more 2D projections is a simple generalization of the described. Parallel projection geometry is assumed and an image plane passing through 3D origin point that coincides with an identified or given reference point in every image is considered; thus, every point and direction found in the image plane can and will be expressed as 3D entities using reference point and known orientation.

Let $L_1(1)$, $L_1(2)$, $L_1(3)$, ... be a sequence of points representing the tubular organ's centerline in the first image and $L_2(1)$, $L_2(2)$, $L_2(3)$, ... be a sequence of points representing the tubular organ's centerline in the second image. Using previous notation, $V_1$ and $V_2$ are projection directions and $V_{12}$ is epi-polar direction orthogonal to two view vectors. Index $i$ is used as an index of a point on line $L_1$ and $j$ as index of point on line $L_2$. Let $R_1(i),R_2(j)$ be corresponding measures of radii from 2 projections, $D_1(i),D_2(j)$ measures of centerline normalized direction vectors and $S_1(i),S_2(j)$ measures of projected cross section areas based on densitometry calculation. Denote $P_1(i)=\text{dot}(L_1(i),V_{12})$, $P_2(j)=\text{dot}(L_2(j),V_{12})$ as epi-polar distances and $E_1(i)=\text{dot}(D_1(i),V_{12})$, $E_2(j)=\text{dot}(D_2(j),V_{12})$. An equivalent definition of E can be given via increments of epi-polar distance and line length $E=dP/dL$.

Consider a function F of two variables $i$ and $j$ defined on the rectangular domain of indexes ij: $(1 \le i \le N) \times (1 \le j \le M)$ where $N$ and $M$ are numbers of points in the centerlines.

$F(i,j) = F_1(|P_1(i)-P_2(j)|) + C_2F_2(|R_1(i)-R_2(j)|) + C_3F_3(|S_1(i)E_2(j)-S_2(j)E_1(i)|) + F_4(E_1(i)E_2(j))$. Here $F_1$, $F_2$, $F_3$, $F_4$ are functions with the following properties. $F_1(0)=F_2(0)=F_3(0)=0$; $F_1$, $F_2$, $F_3$ are monotonically increasing functions; $F_1(\infty)=\infty$; $0 \le F_2, F_3 \le 1$;

$$F_4(x) = \begin{cases} 0, & if \quad x \ge 0 \\ \infty, & if \quad x < 0 \end{cases}$$

$C_2$ and $C_3$ are weight coefficients.

(b) using three or more projection lines to determine a 3D point; for example, a point that minimizes sum of distances from those lines.

Also, in the case of three or more projections, a direction correspondence criterion is incorporated into the optimization process, described above as "defining a novel constraint for the process of 3D reconstruction from three or more views".

If the reference point is one of the skeleton point, i.e. $L_1(i_0)$ and $L_2(j_0)$, an additional constraint forcing the optimization algorithm to path through the reference $i_0, j_0$ is imposed into the target function

$$F_{ref}(i,j) = \begin{cases} \infty, & if \quad (i = i_0 \; and \quad j \neq j_0) \quad or \quad (i \neq i_0 \quad and \quad j = j_0) \\ 0, & otherwise \end{cases}$$

Note that only one term in the target function depends on reference point – the penalty term $F_1(|P_1(i)-P_2(j)|)$.

When the reference point is not known, the difference of epi-polar distances can be described as one parameter family of functions depending on a shift along epi-polar direction. The reference point (or shift) can be found in different ways:

- the reference point can be chosen among the points obeying the condition $E_1(i)=0$ and $E_2(j)=0$, where E is dP/dL, P is epi-polar distance and L is centerline length.
- the shift and therefore reference point can be found via correlation of functions $S_1/E_1$ and $S_2/E_2$ expressed as functions of epi-polar distance to arbitrary temporary reference point or via correlation of functions $R_1$ and $R_2$.
- the reference point can be found in the process of solving the optimization problem if the classical penalty term $F_1(|P_1(i)-P_2(j)|)$ is substituted with the following expression $F_1(|P_1(i)-P_2(j)-(P_1(i_{start})-P_2(j_{start}))|)$ , where $P_1$, $P_2$ are distance to arbitrary temporary reference point and $i_{start}, j_{start}$ are indexes of the first point of the currently optimal matching segment in the point $(i,j)$.

is most informative for the purpose of three-dimensional reconstruction when the viewing direction is orthogonal to the organ. In addition, a pair of projections is more informative when the viewing directions are sufficiently separated. These properties are obviously local, per segment of the organ. Thus, one combination of projections is preferable for one segment of the organ, while another combination of projections is preferable for another.

The third group of embodiments proposes to determine local weights of combination of two 2D image sources for refined 3D reconstruction. Local weights are determined according to the angle between the primary 3D model (reconstructed from the first two projections) centerline and view vectors of the projections and angle between the view vectors.

Accordingly, the third group of embodiments relates to a novel method and system for an automated three-dimensional reconstruction of an organ from three or more projections. Once a three-dimensional reconstruction of the organ from two projections is available, the present group of embodiments provides a method and system that performs an automatic identification of the reconstructed organ in the 2D image of an additional projection, performs an automatic trace and analysis of the organ in the 2D image (in the same manner as was performed on the first and second images), and finally incorporates the new projection into the three-dimensional reconstruction, improving the accuracy of the three-dimensional reconstruction. Such an approach is particularly applicable to imaging an artery contained in an arterial tree.

A projection of an organ is most informative for the purpose of three-dimensional reconstruction when the viewing direction is orthogonal to the organ, and substantially different views produce more accurate 3D reconstruction than not substantially different (close) views. To implement these two notions, some of the embodiments of this third group are directed to a novel method and system for exploiting the three-dimensional reconstruction made from two views to determine local weights for a refined reconstruction incorporating additional projections.

Specifically, the present group of embodiments relates to two aspects of making a better three-dimensional reconstruction of an organ when additional angiographic

Method for three-dimensional reconstruction from N (N>2) 2D projections

Reconstruction of a 3D line from multiple projections can be posed as an optimization problem whose elementary step is reconstruction of a single point. Theoretically, reconstruction of a single point using multiple projections can be done by means of intersection of projective lines corresponding to the 2D projections; in practice, projective lines do not intersect. One natural definition for a 3D reconstructed point, resulting from intersecting two lines, could be defining the 3D point as the middle of the shortest segment connecting the projective lines. Three-dimensional reconstruction from three or more projections expands the above-mentioned idea and determines a 3D point in a similar way. One example is a direct expansion of taking the 3D point that minimizes distance from (three or more) projective lines. Another method is to take the 3D points that result from all pairs of projections and set the final reconstructed point as a geometrical function of these points. The present group of embodiments suggest a novel approach, in which the results from all pairs of projections are indeed used, but rather than setting the 3D reconstructed result as a function of only the points, the relationship between viewing angles and the 3D model is utilized to determine weights for each pair result.

Let $V_1,V_2,\ldots,V_N$ be viewing directions and $L_1,L_2,\ldots,L_N$ be projective lines, $L_i = P_i + \lambda V_i$, where $P_1,P_2,\ldots,P_N$ are points from 2D centerlines (i indexing a projection).

Let $A$ be a 3D model of an organ segment reconstructed from two projections with indexes 1 and 2. We hold, as a result of a primary reconstruction from these two projections 1 and 2, a reference from points $P_1$, $P_2$ to the centerline points of the model $A$. Let $T$ be a local tangent direction of the 3D model $A$ at the region where points $P_1$, $P_2$ reference to. Let $R_{ij}$ be middle of the shortest segment between the projective lines $L_i$, $L_j$. Denote by $W_{ij} = det\ (V_i V_j T)$ a determinant of 3-by-3 matrix composed of unit vectors $V_i V_j$ and $T$. The intersection point is given by the expression:

$$R = \Sigma W_{ij} R_{ij} / \Sigma W_{ij}$$

The quality of intersection is:

$$D = \Sigma W_{ij} D_{ij} / \Sigma W_{ij},$$

projection may rotate the 3D to that projection enabling the operator to appreciate it with comparison to the 2D images (for example).

Color coding of 3D model and/or graphs and other data. Color coding can be implemented to denote narrowing severity, angulation, etc. (or a combination of parameters), to draw the physician attention to problematic segments.

Correlated data. A cross-reference of data from a 2D trace of the vessel to the 3D model to graphs; every point can be allocated simultaneously on all. Cues are presented, for example to enable the operator to investigate the data either specifically or simultaneously.

One or more graphs (see, for example, screen shot, Fig. 34) may be presented including a graph for representing the cross-section area (fusion output) data and one for diameter information, or a combined graph. A diameter data graph may be referred to "eccentricity", as it presents maximum and minimum diameter value for every point along the vessel.

Epi-polar warnings/bars/lines. Epi-polar geometry is well-known and extensively documented, and is used for 3DR in the present invention. However, a 3DR is only as good as the images are to prepare it. Accordingly, to determine whether a second image, in combination with the first image, is adequate to aid in 3DR, embodiments of the present invention provide an operator with a visual indicator. As shown in Figs. 35A-35B, once the operator completes marking of a first image (Fig. 35A), and the vessel of interest is traced, as soon as the operator clicks on/about the stenosis on the second image (Fig. 35B), the system presents, on the second image, epi-lines (lines 3510 and 3520) that are in the vicinity of the first image's markings and epi-bar 3530.

The bar indicates the conditions for 3DR. In the present illustration (Fig. 35B), the bar is color coded to indicate whether the second image is a good combination with the first image. Here, the more "white" the bar is, the better the conditions for 3DR. Accordingly, since the bar in Fig. 35B is quite white, conditions are good for 3DR (a "redder" bar would indicate poorer conditions for 3DR).

WHAT IS CLAIMED IS:

1. A method for three-dimensional reconstruction (3DR) of a single tubular organ using a plurality of two-dimensional images comprising:

displaying a first image of a vascular network;

receiving input for identifying on the first image a vessel of interest;

tracing the edges of the vessel of interest including eliminating false edges of objects visually adjacent to the vessel of interest;

determining substantially precise radius and densitometry values along the vessel;

displaying at least a second image of the vascular network;

receiving input for identifying on the second image the vessel of interest;

tracing the edges of the vessel of interest in the second image, including eliminating false edges of objects visually adjacent to the vessel of interest;

determining substantially precise radius and densitometry values along the vessel of interest in the second image;

determining a three dimensional reconstruction of the vessel of interest; and

determining fused area measurements along the vessel.

2. The method according to claim 1, wherein the vessel of interest is selected from the group comprising: an artery, a vein, a coronary artery, a carotid artery, a pulmonary artery, a renal artery, a hepatic artery, a femoral artery, a mesenteric artery, and any other tubular organ.

3. The method according to claim 1, further comprising determining a centerline, comprising a plurality of centerline points.

10. The method according to claim 1, wherein elimination of false edges comprises ignoring one or more bubbles adjacent the vessel of interest.

11. The method according to claim 1 or 10, wherein elimination of false edges comprises:

defining a region of interest substantially parallel to a primary centerline;

detecting at least one cluster of pixel data, adjacent to the vessel of interest, wherein each cluster of pixel data having a predetermined brightness level greater than a brightness level of surrounding pixel data;

selecting an arbitrary pixel within each cluster;

selecting a second pixel provided on a lane bounding the region of interest for each arbitrary pixel of each cluster;

establishing a barrier line to define an edge for the vessel of interest by connecting a plurality of arbitrary pixels with a corresponding second pixel, wherein upon the tracing each edge of the vessel of interest, the traced edge avoids each barrier line.

12. The method according to claim 1, wherein elimination of false edges comprises detecting and/or eliminating one or more bumps along the vessel of interest.

13. The method according to claim 1 or 12, wherein elimination of false edges includes:

establishing a list of suspect points, comprising:

establishing a plurality of first distances between each of a plurality of originating points on at least one preliminary traced edge and a corresponding closest point positioned along the primary centerline;

dividing each edge into a plurality of segments using the anchor points, wherein for each segment, correspondence between the edges is established in that every point of each edge includes at least one pair of points on an opposite edge and a total sum of distances between adjacent points is minimal; and

connecting the centers of the plurality of segments to determine the centerline.

15. The method according to claim 1, wherein determining densitometry values comprises subtracting a background influence.

16. The method according to claim 1 or 15, wherein determining densitometry values comprises:

establishing a plurality of profile lines substantially parallel to at least one edge of the vessel of interest;

establishing a parametric grid covering the vessel of interest and a neighboring region, wherein the parametric grid includes a first parameter of the vessel of interest along the length thereof and a second parameter for controlling a cross-wise change of the vessel of interest;

sampling the image using the grid to obtain a plurality of corresponding gray values, wherein:

the gray values are investigated as functions on the profile lines;

substantially eliminating detected occluding structures on the outside of the vessel of interest, the structures being detected as prominent minima of the parameters;

re-determining $p_2$ substantially in accordance with $p_{2new} = p_2 + \delta$, where $\delta$ is a smooth compensatory function establishing correspondence of one or more landmark points.

22. The method according to claim 1, further comprising displaying color coded data relating to the vessel of interest in any display of data.

23. The method according to claim 1, wherein after receiving input for identifying the vessel of interest in the second image, displaying an epi-polar indicator for indicating a concurrence between the first image and second image for producing a three-dimensional reconstruction of the vessel of interest.

24. The method according to claim 1, further comprising displaying quantitative analysis of the vessel of interest including cross-section area graph and/or lesion analysis measurements.

25. The method according to claim 1, further comprising cross-referencing data among at least a pair or more data related to the two-dimensional trace of the vessel of interest, the three-dimensional reconstruction of the vessel of interest, and graphical data.

26. A system for three-dimensional reconstruction (3DR) of a single blood vessel using a plurality of two-dimensional images comprising:

a display for displaying a first image of a vascular network and a second image of a vascular network, and a three-dimensional reconstruction of a vessel;

input means for receiving input for identifying a vessel of interest on the first image and for identifying the vessel of interest on the second image;

a processor arranged to operate one or more application programs comprising computer instructions for:

tracing the edges of the vessel of interest including eliminating false edges of objects visually adjacent to the vessel of interest;

fusing a majority of the data into single area measurements, weighting each source of data according to the reliability of the data.

31. The system according to claim 30, where weighting is computed as a function of the views geometry and/or 3D vessel geometry.

32. The system according to claim 26, wherein the input for identifying the vessel of interest comprises of three points comprise a first point to mark the stenosis general location, a second point proximal to the stenosis, and a third point distal to the stenosis.

33. The system according to claim 26, wherein the input for identifying the vessel of interest comprises markers for two (2) points for at least one of the first and second images, wherein one of the two points is anywhere proximal to the stenosis and the other point is anywhere distal to the stenosis.

34. The system according to claim 26, wherein the markers comprise two (2) points for the first image and one (1) point for the second image, wherein one of the two points is anywhere proximal to the stenosis and the other point is anywhere distal to the stenosis and wherein one point is an anchor point identified automatically on the first image.

35. The system according to claim 26, wherein elimination of false edges comprises ignoring one or more bubbles adjacent the vessel of interest.

36. The system according to claim 26 or 35, wherein elimination of false edges comprises:

defining a region of interest substantially parallel to a primary centerline;

detecting at least one cluster of pixel data, adjacent to the vessel of interest, wherein each cluster of pixel data having a predetermined brightness level greater than a brightness level of surrounding pixel data;

being greater than a predefined value, the corresponding edge point is determined to be a bump point in a bump;

determining a bump area defined by a plurality of connected bump points and a cutting line adjacent the vessel of interest, wherein the cutting line comprises a line which substantially maximizes a ratio between the bump area and a power of a cutting line length; and

cutting the bump from the edge at the cutting line to establish a final edge.

39. The system according to claim 26, wherein the application programs also include computer instructions for displaying an epi-polar indicator for indicating a concurrence between the first image and second image for producing a three-dimensional reconstruction of the vessel of interest.

40. The system according to claim 28, wherein defining a centerline of the vessel of interest comprises:

determining final traced edges of the vessel of interest;

determining pairs of anchor points, wherein each pair comprises one point on each edge;

determining a cross-sectional line by searching for pairs of anchor points which, when connected, establish the cross-sectional line substantially orthogonal to the center-line;

dividing each edge into a plurality of segments using the anchor points, wherein for each segment, correspondence between the edges is established in that every point of each edge includes at least one pair of points on an opposite edge and a total sum of distances between adjacent points is minimal; and

determining cross-sectional area using the eliminated prominent minima.

43. The system according to claim 26, further comprising determining healthy vessel dimensions using an iterative regression over a healthy portion of the vessel of interest.

44. The system according to claim 43, wherein each iteration comprises a compromise between a pre-defined slope and a line that follows healthy data.

45. The system according to claim 44, wherein the compromise is toward the line that follows the healthy data if the line corresponds to actual data over a plurality of clusters.

46. The system according to claim 28, wherein determining the three-dimensional reconstruction of the vessel of interest includes:

determining a conventional epi-polar distance $p_1$ for the plurality of centerline points in the first image;

determining a conventional epi-polar distance $p_2$ for the plurality of centerline points in the second image; and

re-determining $p_2$ substantially in accordance with $p_{2new} = p_2 + \delta$, where $\delta$ is a smooth compensatory function establishing correspondence of one or more landmark points.

47. The system according to claim 26, wherein the application programs including computer instructions for displaying color coded data relating to the vessel of interest in any display of data.

48. The system according to claim 26, further comprising epi-polar indicator means for indicating a concurrence between the first image and second image for producing a three-dimensional reconstruction of the vessel of interest.

extracting centerlines of the tubular organ on two angiographic images;

obtaining invariant functions of the two images;

constructing an optimization target function that is comprised of a penalty function expressing soft epi-polar constraint and discrepancies between invariant functions; the optimization's target function is defined over all possible correspondences between the two centerline points;

solving the optimization target function, to generate a map between 2D points on one centerline to the 2D points on the other centerline;

if a reference point is given, then optimizing solution so that the map includes the match of the reference point;

when a reference point is not given, finding it either by obeying the condition $E_1(i)=0$ and $E_2(j)=0$ where E is dP/dL, P is epi-polar distance and L is centerline length, or by means of finding the correlation of functions $S_1/E_1$ and $S_2/E_2$ expressed as functions of epi-polar distance to arbitrary temporary reference point or via correlation of functions $R_1$ and $R_2$;

wherein every matched set of 2D points defines a 3D point, for example as a point that minimizes distance from projective lines and the sequence of these 3D points is the three-dimensional reconstruction of the tubular organ.

53. The method according to claim 52, wherein invariant functions are comprised of radius or projected cross section area or centerline direction of the tubular organ along the centerline points, or the invariant function obtaining an invariant function from a tubular organ characteristic, the tubular organ being imaged in angiography, the invariant function being equivalence between the ratio of projected area and the visible epi-polar orientation for every pair of views..

determining a shift between the third angiographic projection and the projected 3D reconstruction on said image plane so as to identify the tubular organ within the third angiographic projection;

tracing and analyzing the tubular organ in the third angiographic projection using the projected 3D reconstruction on said image plane as a first approximation so as to derive properties of the tubular organ; and

using said properties for re-determining the three-dimensional reconstruction to a better approximation.

61. The method according to claim 60, wherein projecting the 3D reconstruction onto an image plane produces a binary projected image.

62. The method according to claim 60, wherein projecting the 3D reconstruction onto an image plane produces a realistic projected image, in which a pixel's gray-level is a function of the length of intersection between the ray and the model.

63. A method for three-dimensional reconstruction from N (N>2) 2D projections, comprising:

obtaining a three-dimensional reconstruction for every pair of projections;

assigning a respective weight per 3D point, for each of said pairs of projections, that reflects a mutual geometry of two views and local orientation of a primary 3D model in such a way that maximal weight (1) is achieved by the combination of two orthogonal views, which are also both orthogonal to the organ; and such that the respective weight is near zero in the case when the two views are close to each other or if one of the views is too oblique; and

defining the reconstructed 3D point as a weighted sum of the intersection points per each pair of projective lines.

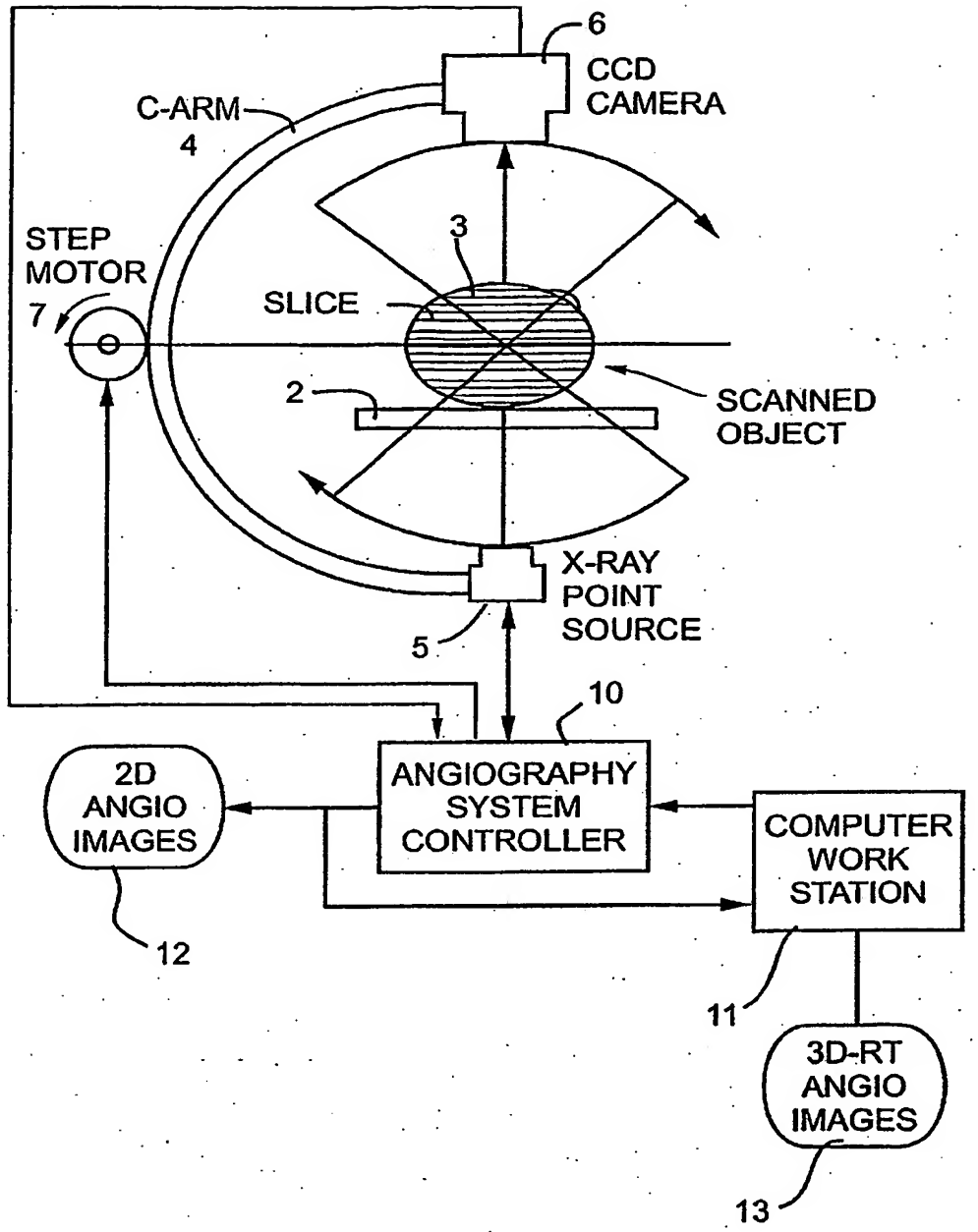

6
CCD
CAMERA
C-ARM
4
3
STEP
MOTOR
7
SLICE
2
SCANNED
OBJECT
X-RAY
POINT
SOURCE
5
10
2D
ANGIO
IMAGES
ANGIOGRAPHY
SYSTEM
CONTROLLER
COMPUTER
WORK
STATION
12
11
3D-RT
ANGIO
IMAGES
13

Fig. 1

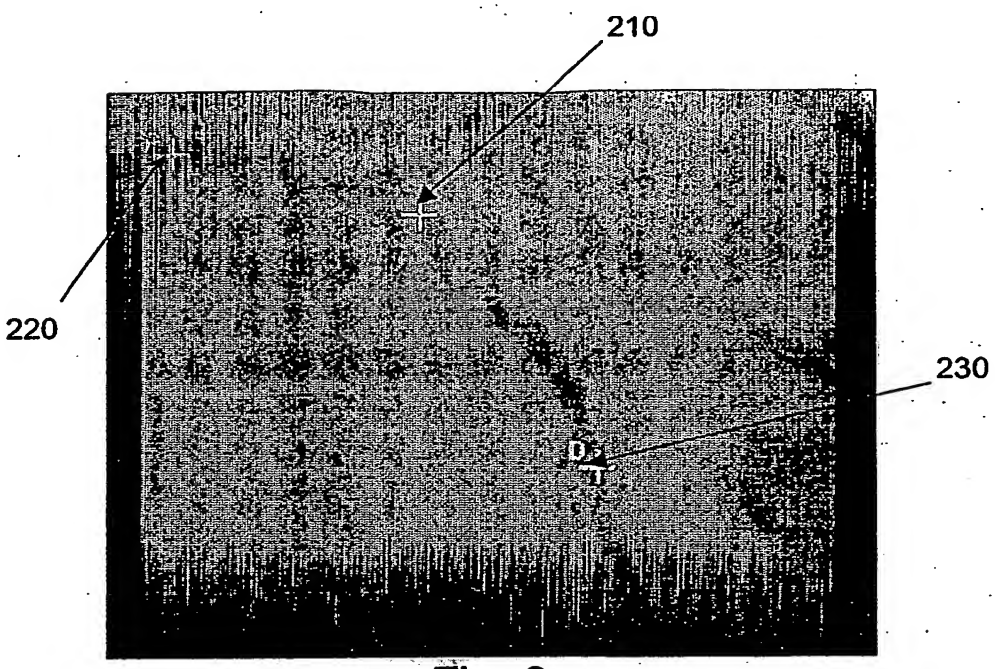



Fig. 2

Fig. 3


Fig. 4

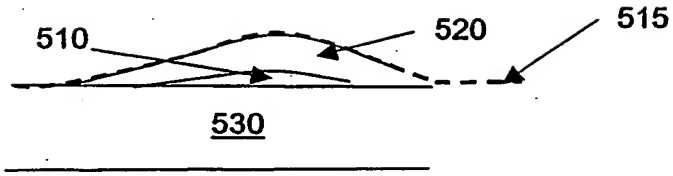



Fig. 5

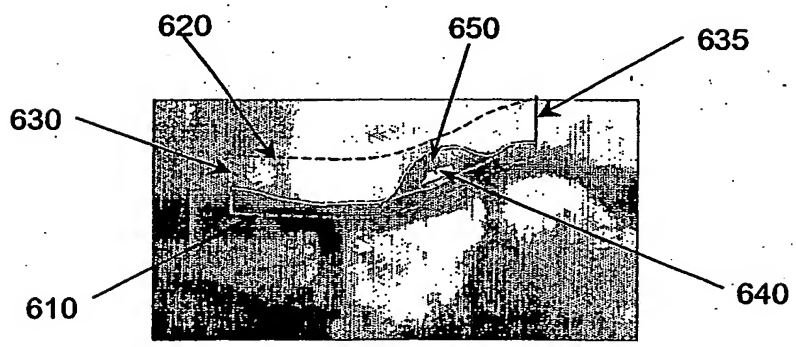



Fig. 6


Fig. 7


Fig. 8


Fig. 9


Fig. 10

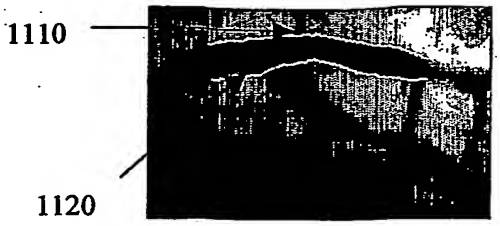



Fig. 11

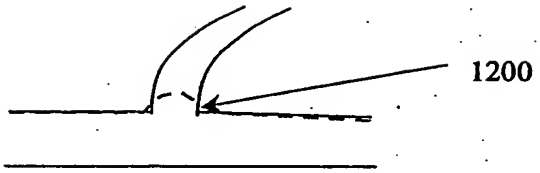



Fig. 12

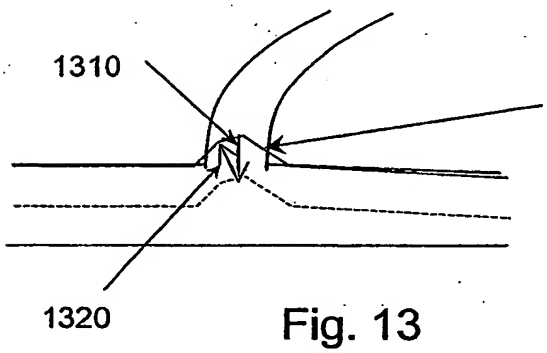



Fig. 13

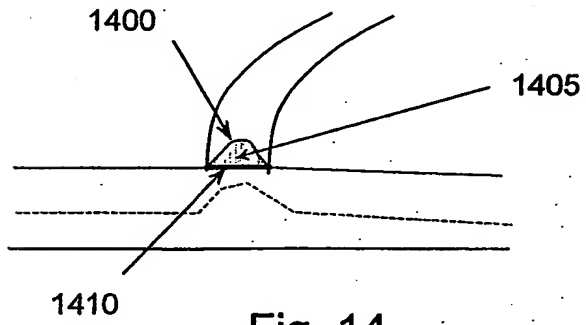



Fig. 14

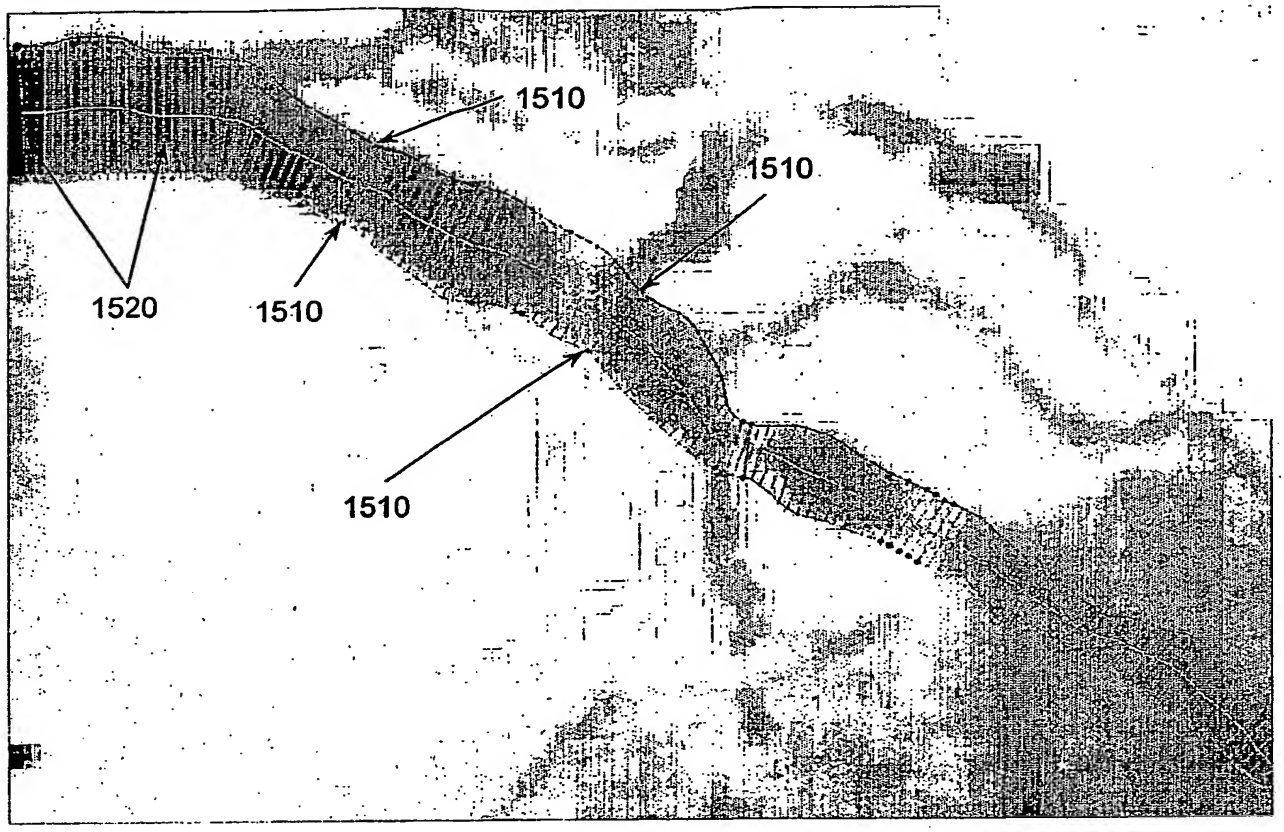
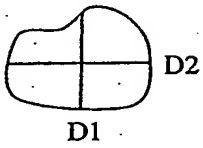



Fig. 15

Fig. 16



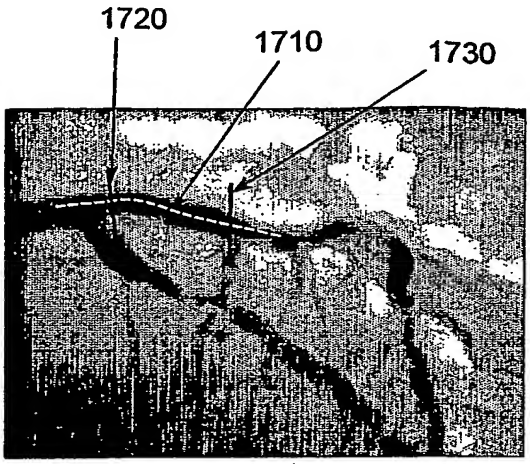



Fig. 17

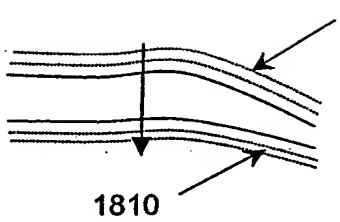



Fig. 18

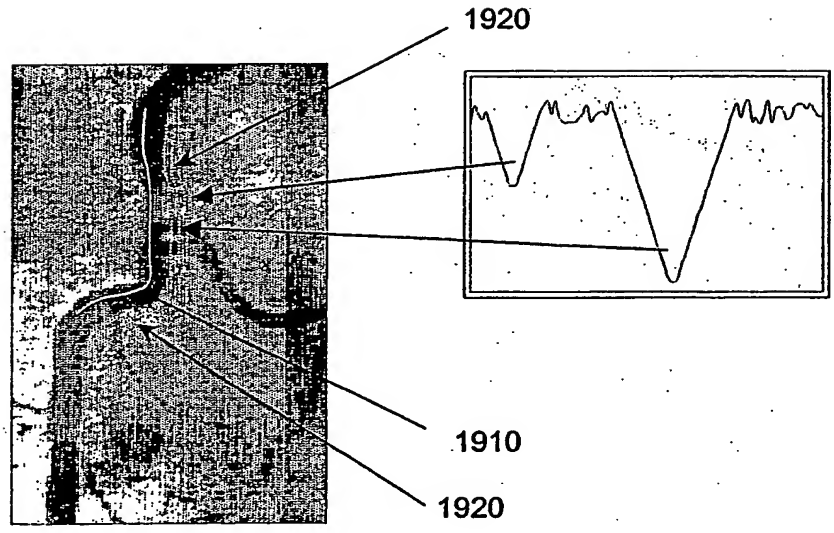



Fig. 19A

Fig. 19B

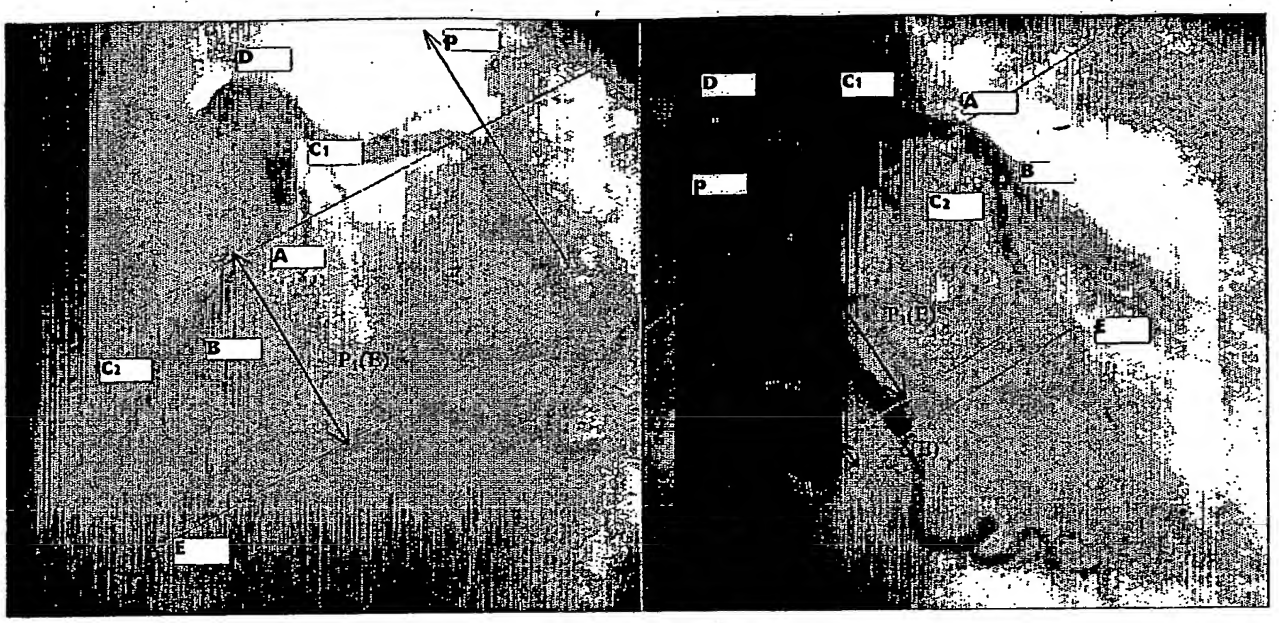



Fig. 20A

Fig. 20B

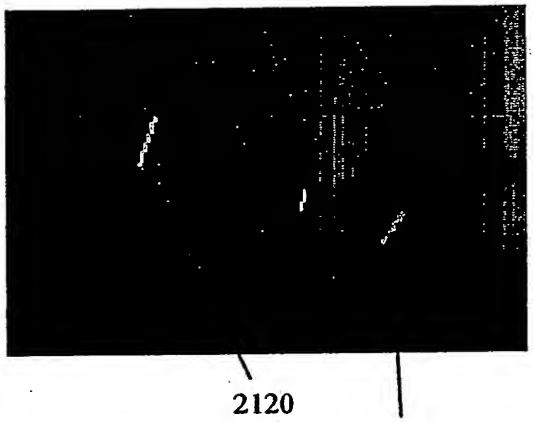



Fig. 21A

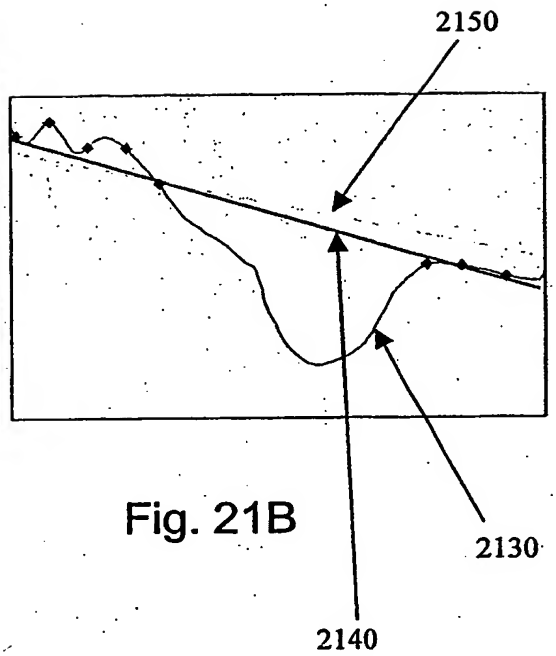



Fig. 21B

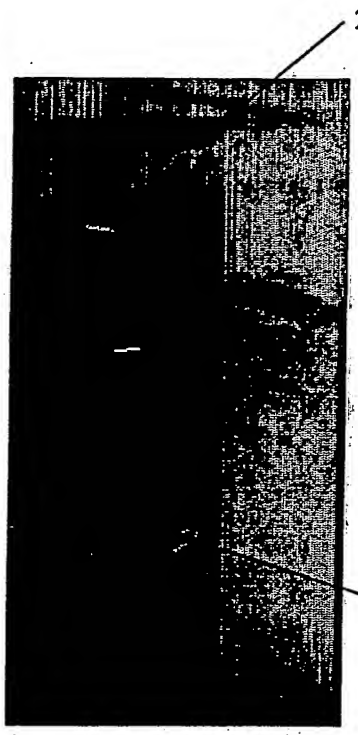



Fig. 22A

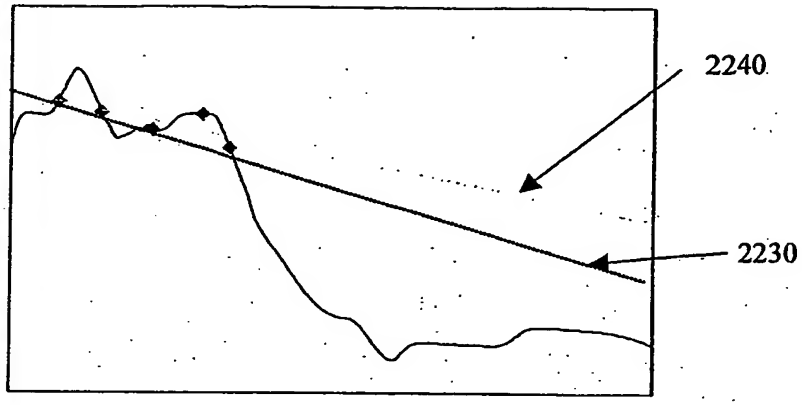



Fig. 22B

Fig. 23

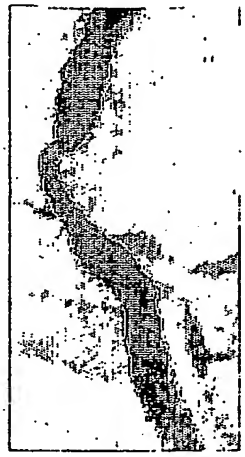


Fig. 24

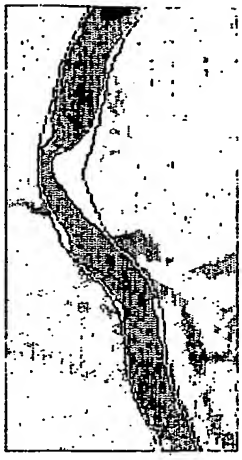


Fig. 25

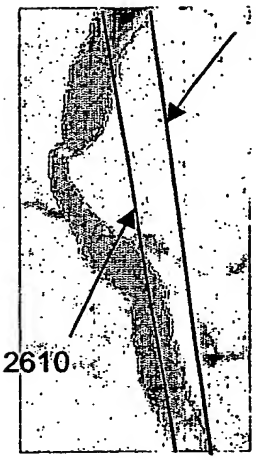



Fig. 26


Fig. 27

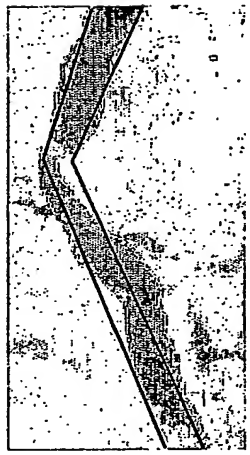


Fig. 28

## 2D Display Screenshot

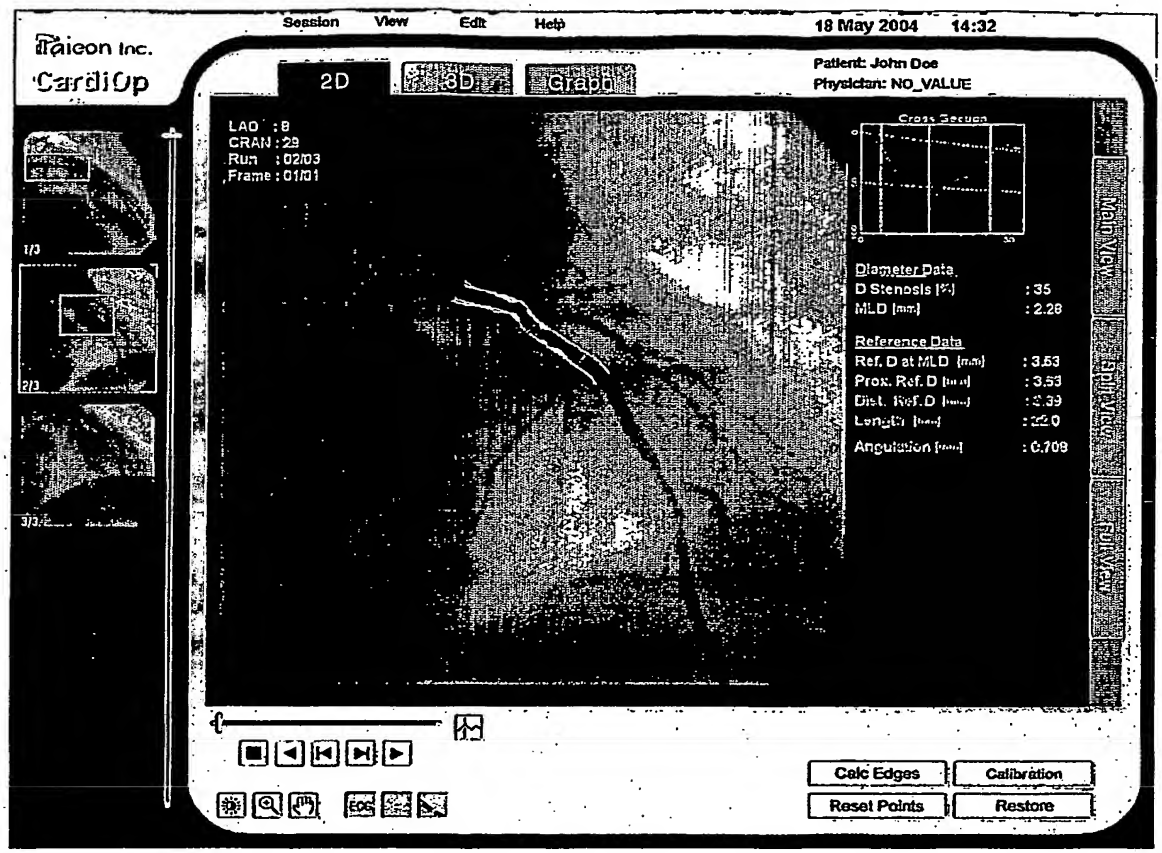



Fig. 29

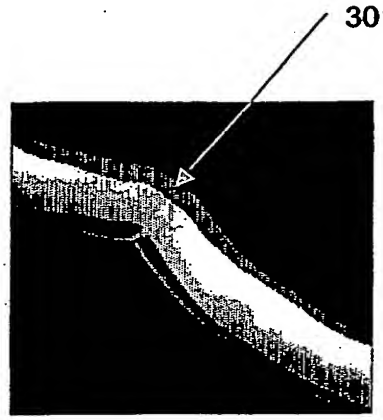



Fig. 30

3DR Display - Screenshot

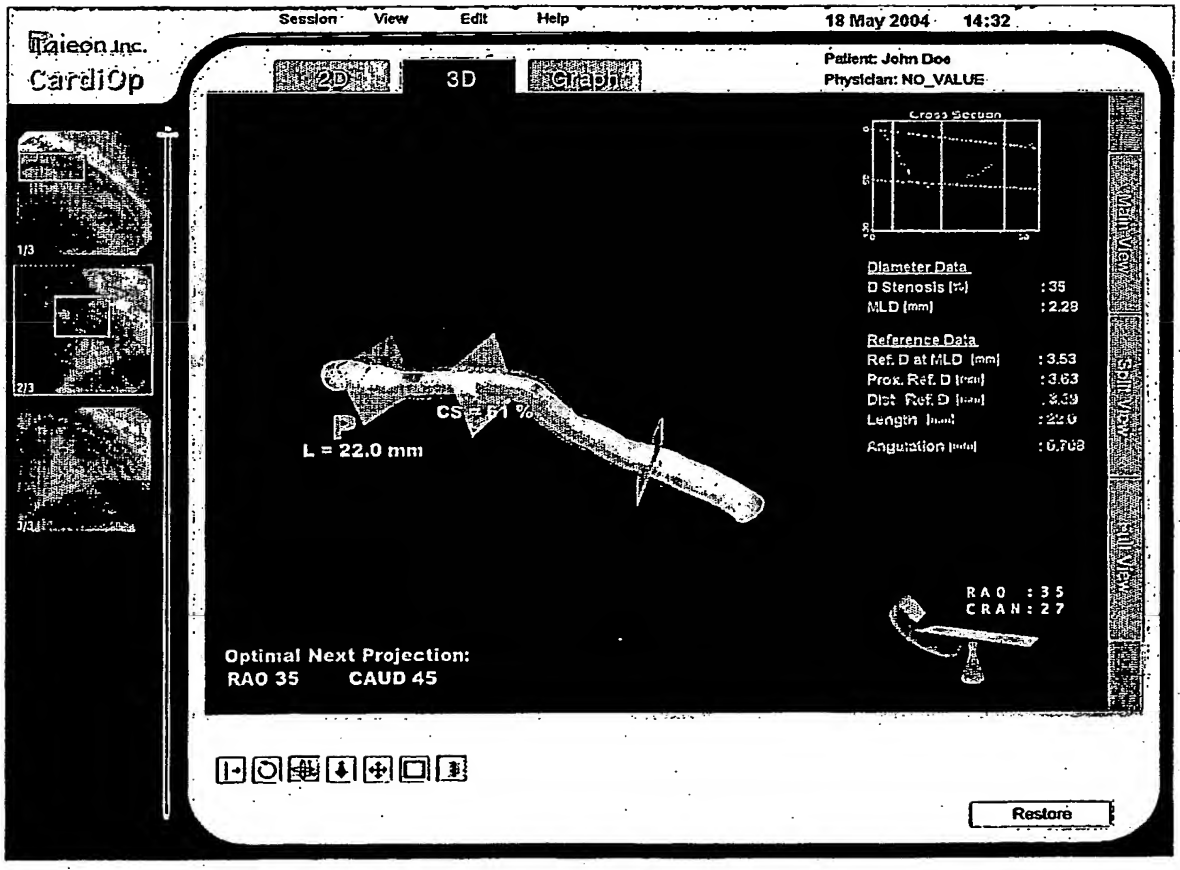



Fig. 31

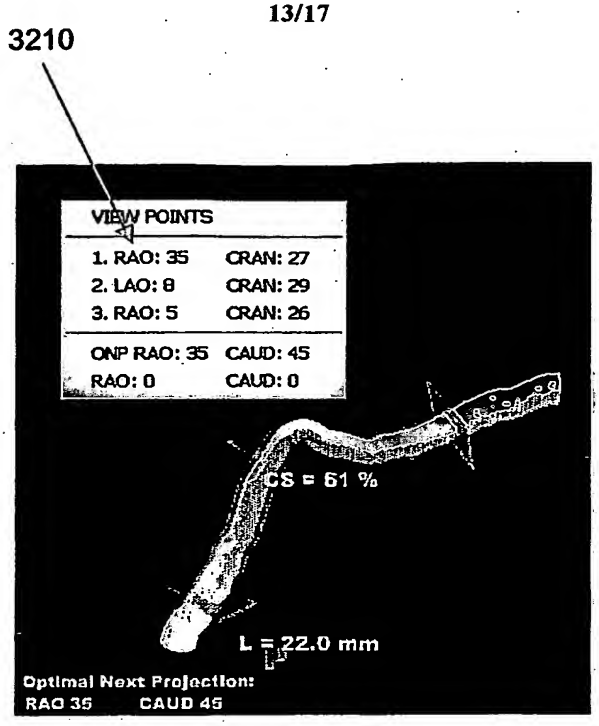



Fig. 32

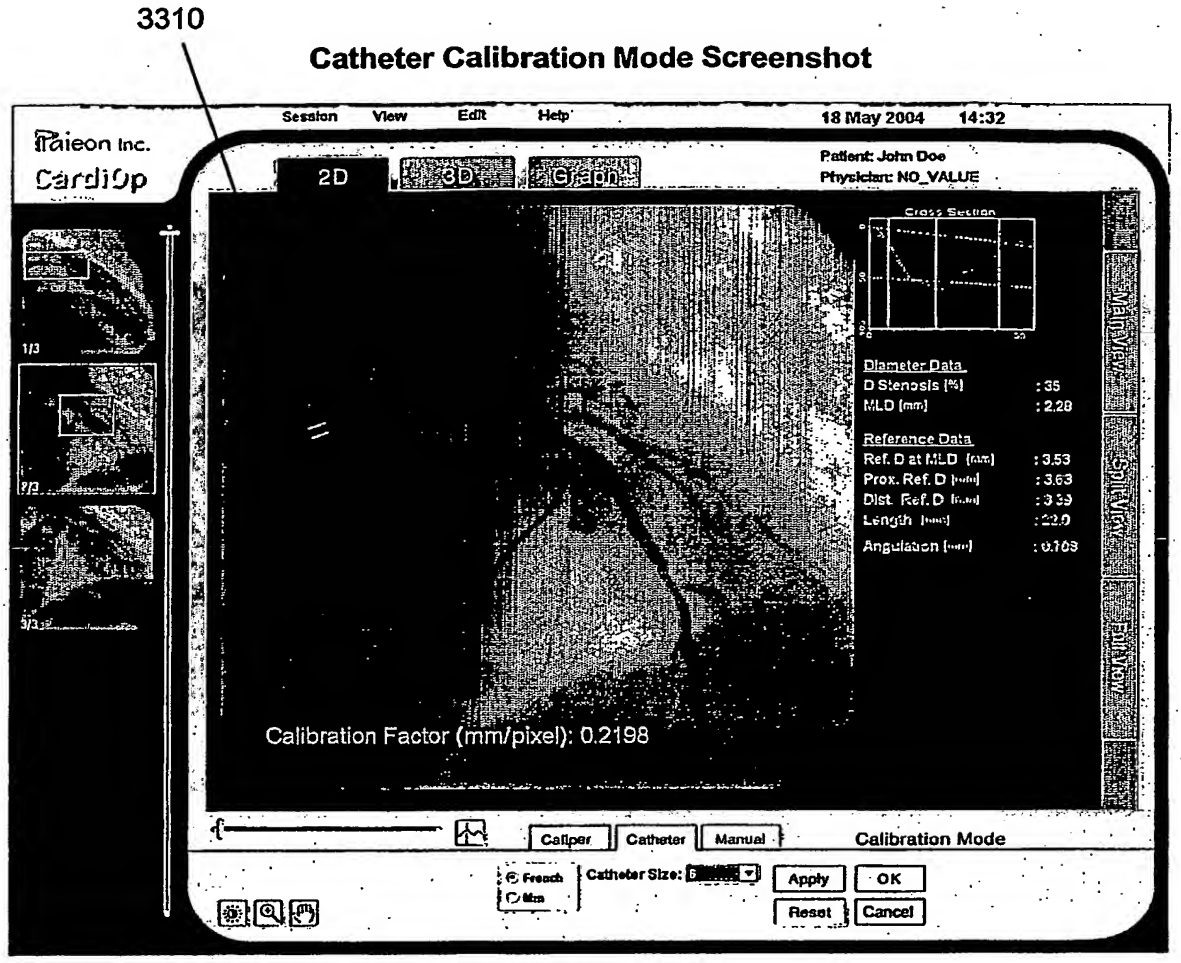



Fig. 33

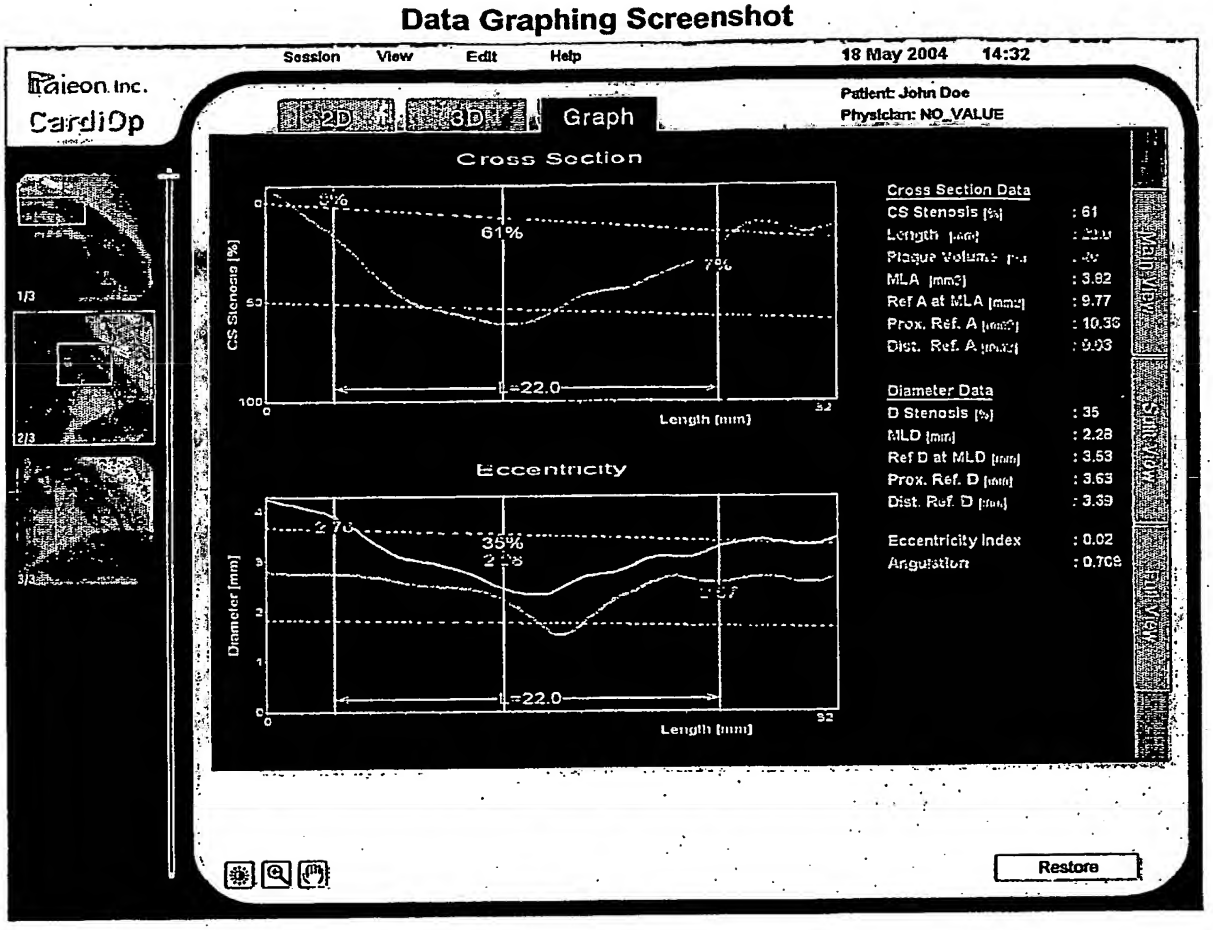



Fig. 34


Fig. 35A


Fig. 35B

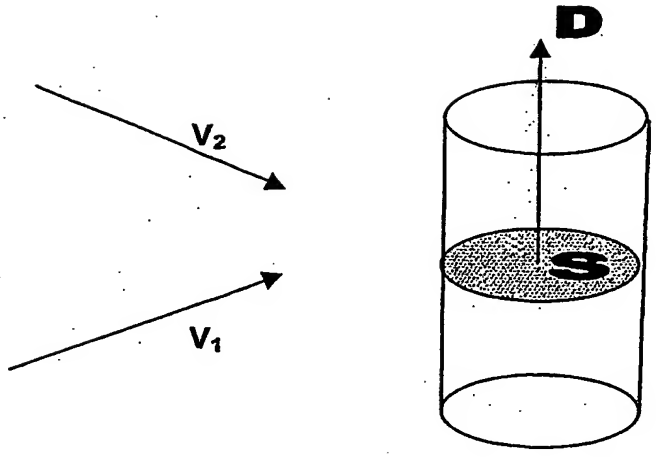



FIG. 36

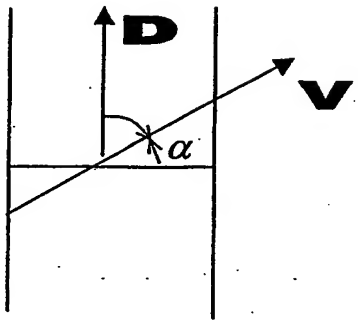



FIG. 37